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Dynamic input to determine hip joint moments, power and work on the prosthetic limb of transfemoral amputees: ground reactions *vs* knee reactions

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ABSTRACT

Study Design: Comparative analysis

Background: Calculations of lower limbs kinetics are limited by floor-mounted force-plates.

Objectives: Comparison of hip joint moments, power and mechanical work on the prosthetic limb of a transfemoral amputee calculated by inverse dynamics using either the ground reactions (force-plates) or knee reactions (transducer).

Methods: Kinematics, ground reactions and knee reactions were collected using a motion analysis system, two force-plates and a multi-axial transducer mounted below the socket, respectively.

Results: The inverse dynamics using ground reactions under-estimated the peaks of hip energy generation and absorption occurring at 63 % and 76 % of the gait cycle (GC) by 28 % and 54 %, respectively. This method over-estimated a phase of negative work at the hip (from 37 %GC to 56 %GC) by 24%. It under-estimated the phases of positive (from 57 %GC to 72 %GC) and negative (from 73 %GC to 98 %GC) work at the hip by 11 % and 58%, respectively.

Conclusions: A transducer mounted within the prosthesis has the capacity to provide more realistic kinetics of the prosthetic limb because it enables assessment of multiple consecutive steps and a wide range of activities without issues of foot placement on force-plates.

CLINICAL RELEVANCE

The hip is the only joint that an amputee controls directly to set in motion the prosthesis. Hip joint kinetics are associated with joint degeneration, low back pain, risks of fall, etc. Therefore, realistic assessment of hip kinetics over multiple gait cycles and a wide range of activities is essential.

KEYWORDS

Inverse dynamics ; Ground reactions ; Direct measurements ; Hip kinetics ; Transfemoral amputation

1. INTRODUCTION

Net joint forces and moments as well as power and work at the ankle, knee and hip of the sound (1) and prosthetic limbs of transfemoral amputees have been used to determine the effects of rehabilitation programs (2-3), alignments of prosthesis (4-6) and prosthetic components (7-11).

1.1. Inverse dynamics

The most comprehensive method to compute these variables relies on well-established inverse dynamic equations. However, several studies demonstrated that this calculation method presents some shortcomings (12-15). It is based on assumptions that rigid segments are linked by ideal joints. It is sensitive to input data such as inertial parameters, time derivatives, location of centre of pressure and joint centre thought external markers. Finally, in principle, errors are increasingly propagated upward between the ankle, the knee and the hip (15).

1.2. Limitations floor-mounted force-plates

By definition, the inverse dynamics method required kinematics and ground reactions forces that are typically obtained using a 3D motion capture system and force-plates, respectively. Most of the limitations associated with the calculation of lower limbs kinetics are inherent to the experimental setting of these instruments, particularly the floor-mounted force-plates.

As indicated by Favre et al (2010) "*mainly due to the price of the motion capture systems, standard gait laboratories have the capability to measure only a few consecutive steps of ground walking*"[(16)p-2196]. In addition, the sole contact of each foot on a force-plate can be achieved through personalized arrangements of walking start point and/or the position of force-plates to avoid targeting and/or repetitive recording of invalid trials. The number of steps analysed depends on the number of force-plates. Some of these limitations can be alleviated using an instrumented treadmill. Nonetheless, relying on anchored equipments is often perceived as a significant technical constraint, particularly when assessing activities of daily living (e.g., ascending and descending stairs and slopes) of lower limb amputees who rely less on proprioception for foot placement. Thus, the measurements might be only partially reflective of a natural gait.

1.3. Benefits of using direct measurement of knee reactions

Several studies have presented an alternative method that can potentially alleviate the limitations mentioned above. In these cases, a multi-axial transducer mounted below the residuum of transfemoral amputees fitted with a socket or osseointegrated fixation was used to measure directly the forces and moments applied on the residuum and the prosthetic knee [(17-19)]. This apparatus enabled load measurements during of a large number of gait cycles and activities such as ascending and descending stairs and slopes [(20-21)].

However, the use of a transducer to validate inverse dynamics results has received little interests. Only one study has recently compared the forces and moments applied on the prosthetic knee of a transfemoral amputee during walking that were measured directly by a transducer with the ones calculated by three inverse dynamics computations corresponding to three and two segments, and ground reaction vector technique[(22)]. The results demonstrated that the differences between the calculated and measured forces and moments were relatively small. However, this study also showed that the dynamic outcomes of the prosthetic components (i.e., absorption of the foot, friction and limit stop of the knee) were only partially assessed with inverse dynamic methods. This information is critical to assess all aspects related to the construction and, eventually, usage of the prosthesis.

1.4. Need for knee reactions to determine hip joint kinetic

The use of a transducer to improve implementation of inverse dynamics equations is yet to be explored, particularly the possibility to use the knee reactions measured by a transducer as dynamic input to determine of hip joint kinetics. Indeed, there is a need for an additional comparative study focusing on all aspects of hip joint kinetics (e.g., moments, power and mechanical work).

Such study is critical because the hip is the only joint that the amputee controls directly to set in motion the prosthesis, particularly during the swing phase. Also, the dynamic outcomes of the prosthesis might have a critical impact on the hip. Therefore, in principle, it could provide further clinical information as hip joint kinetic are potentially associated with joint degeneration, low back pain, risks of fall, etc.

1.5. Purpose

The purpose of the present case study was to compare hip joint moments, power and mechanical work on prosthetic limb of a transfemoral amputee calculated by inverse dynamics, with either the ground reactions obtained from fixed force-plates, or knee reactions directly measured with a transducer, used as input data.

2. METHODS

2.1. Participant

One female transfemoral amputee (36 yr, 1.6 m, 62.6 kg) participated in this study. She provided informed written consent. The research institution's human ethics committee approved this study.

2.2. Apparatus

Figure 1 presents a schematic representation of the components of the prosthetic leg, the marker set and alignment, the position and orientation of the socket and the transducer coordinate systems, and angle correction to re-align both coordinates systems.

The participant walked with a prosthesis including an ischial containment socket, a multi-axial transducer (JR3 Inc., Woodland, CA, USA), and her usual knee (Safety knee, Otto-Bock, Vienna, Austria), a solid ankle cushioned heel (SACH foot, Otto-Bock, Vienna, Austria) and footwear. The socket used was specifically manufactured to replicate the internal geometry of the subject's current socket and to incorporate an adapter attaching the transducer.

Kinematics and ground reactions were recorded simultaneously with a 6-camera Peak-Motus (VICON, Oxford, UK) and two force-plates (AMTI, Watertown, MA, USA) at 50 Hz and 500 Hz, respectively. Double-sided sticky tape was used to place markers on the hip, socket, transducer, pylon and shoe, approximately at landmark levels (i.e., greater trochanter, tibial tuberosity, calcaneum, 5th metatarsal head) and on mechanical parts (i.e., knee axis, ankle fixation).

Knee reactions data was measured and recorded at 200 Hz using a multi-axial transducer, similar to the one used in previous studies [17-18, 20-23], and a laptop, respectively. The three components of forces and moments were measured with accuracy better than 1 N and 1 Nm, respectively. The transducer was attached to the socket using a custom-made spherical plate and to the knee using a pyramidal connector. The transducer was mounted in such a way that the orientation of its coordinate system ($T[X_T, Y_T, Z_T]$) was roughly aligned with the socket coordinate system ($S[ML_S, AP_S, VT_S]$) corresponding to the local anatomical axes of the residuum (ML: Medio-Lateral, AP: Antero-Posterior, VT: Long). The axes of the residuum were determined as the half point between the rims of the socket in the three planes. Both coordinate systems were re-aligned thanks to a transform matrix applied afterward that was created from bench top measurements.

*** Insert Figure 1 here ***

2.3. Recording

First, the prosthetic leg including the transducer was set up and aligned by a qualified prosthetist. The alignment replicated closely the usual one. Approximately 15 min of practice with the instrumented prosthetic leg was allowed before recording to ensure participant confidence, safety and comfort. Then, the participant performed 10 trials of walking along a 5-metre walkway. The participant was asked to step onto the two force-plates while walking at self-selected comfortable speed. Sufficient rest was given between trials to avoid fatigue. Finally, the participant doffed the socket, freeing the prosthesis to allow bench top measurement for the calibration (i.e., zero-offset), re-alignment of coordinate systems and determination of inertial parameters of the prosthetic components.

2.4. Data processing

The raw data were imported into a customized Matlab software program (Math Works Inc, Natick, MA, USA) written to implement the following data processing.

A calibration matrix was applied for each force-plate and the transducer to eliminate cross-talk and correct the offset of electrical zero. The kinematic data was filtered with a classical 4th order Butterworth filter at 5 Hz cut-off frequency. Kinematics and forces-plates data were synchronised with transducer data using the first heel contact on the force-plate as the instant of reference. The inertial parameters of the prosthetic limb were estimated using volume of residuum and bench top measurements of each component [22]. The flexion-extension of knee prosthesis was estimated functionally using SARA method as described in Ehrig et al (2007)[24]. The hip joint centre corresponded to the intersection between the line going from the marker on the greater trochanter to the direction of the flexion-extension axis and the long axis of the residuum.

The forces and moments at the hip joint centre were computed for each gait cycle by 3D inverse

dynamics [25] using the ground reactions at the centre of pressure level and knee reactions at the joint centre level as input data. The former involved two segments: foot and leg, and thigh [22]. The hip joint moments were expressed in frontal, axial and sagittal planes in the pelvis coordinate system, representing the “Flexion-Extension”, “Internal-External rotation” and “Adduction-Abduction”, respectively.
The 3D power was obtained by computing the dot product of net moment and joint angular velocity vectors, both expressed in the pelvis coordinate system. The mechanical work was the area under the curve of power. Positive and negative values corresponded to generation and absorption of energy, and positive and negative work, respectively.
All the curves of moments and power of each gait cycle were time-rescaled from zero to 100 to facilitate averaging of all trials and reporting of events in percentage of gait cycle.

2.5. Data analysis

The overall analysis relied on the average difference, one standard deviation and root mean square error (RMSE) between the mean curves of hip moments and power obtained with both inputs during the full gait cycle as well as support and swing phases. The phases analysis compared time and value of peaks of hip power and total work obtained with both inputs during the typical phases of generation (i.e., H1, H3) and absorption (i.e., H2, H4) [9-10, 26-27]. The comparison relied on the average and one standard deviation of differences between both inputs (the ground reactions minus knee reactions) for each trial, expressed in units or percentage. A positive and negative value corresponded to an under-estimation and over-estimation, respectively, of the results obtained with ground reactions.

3. RESULTS

A total of eight trials were considered for analysis. Two trials were discarded during post analysis of the force-plate data showing that the sole contact of a foot onto one plate was doubtful. The participant walked at 1.02±0.01 m/s with a cadence of 48.79±0.61 steps/min. The support and swing phases represented 60.00±1.20 % and 40.00±1.20 % of the gait cycle, respectively, lasting 1.23±0.02 s.

3.1. Moments

The mean hip joint moments are presented in Figure 2. The average differences in mean moments between both inputs on the frontal, axial and sagittal planes of the pelvis were -0.86±2.59 Nm, 0.97±2.42 Nm and 0.91±3.57 Nm during the full cycle, -0.65±3.30 Nm, 2.19±2.45 Nm and 1.43±3.96 Nm during the support, and -1.17±0.60 Nm, -0.86±0.27 Nm and 0.12±2.74 Nm during the swing, respectively. RMSEs on the frontal, axial and sagittal planes of the pelvis were 2.71 Nm, 2.59 Nm and 3.66 Nm during the full cycle, 3.34 Nm, 3.26 and 4.18 Nm during the support, and 1.31 Nm, 0.90 Nm and 2.71 Nm during the swing, respectively.

*** Insert Figure 2 here ***

3.2. Power

Mean curves of 3D hip power are presented in Figure 3. The average differences in mean power between both inputs were -1.59±5.72 W, -2.51±5.15 W and -0.2±6.3 W during the cycle, support and swing phases, respectively. RMSEs on the power were 5.91 W, 5.69 W and 6.23 W during the cycle, support and swing phases, respectively. Times of occurrence and peak values of the power during each period of generation and absorption are presented in Table 1. The average RMSEs between the two inputs during the phases H1, H2, H3 and H4 were 4.05±0.60 W, 8.82±1.79 W, 9.84±1.77 W and 3.65±0.57 W, respectively. The peak of generation H1 obtained with inverse dynamics using ground reactions was under-estimated by 3.22±2.29 % for four trials and over-estimated by 9.45±10.37 % for four trials. The peak of absorption H2 was under-estimated by 5.70±1.23 % for five trials and over-estimated by 9.44±7.71 % for three trials. The peaks H3 and H4 of generation and absorption were consistently under-estimated by 27.78±9.53 % and by 53.66±6.21 %, respectively.

*** Insert Figure 3 here ***

3.3. Work

The mean work produced during each phasis of power is presented in Table 1. The positive work during the phase H1 determined by inverse dynamics using ground reactions was under-estimated by 3.98±2.54 % for five trials and over-estimated by 4.95±4.61 % for three trials. The negative work during the phases H2 and H4 were consistently over-estimated and under-estimated by 23.37±7.50 % and 58.08±9.28 %, respectively. The positive work during the phase H3 was consistently under-estimated by 11.08±0.31 %.

*** Insert Table 1 here ***

4. DISCUSSION

4.1. Limitations

The scope of this case study was to compare the hip joint kinetics obtained with ground reactions (force-plates) and knee reactions (transducer). By definition, the interpretation and transfer of the results to other transfemoral amputees must be conducted with care mainly because of the typical intrinsic limitations associated with a technical note focusing on a single-case. Only one gait cycle of the prosthetic limb per trial could be assessed given the number of force-plates. The overall results were based on a total of eight cycles.

4.2. Variability

The cycle-to-cycle variability of the participant was low. This confirms results of several studies focusing on intra-participants variability [20-21]. In particular, the variation between the duration of all gait cycles was only 0.05 s. Therefore, the effect of time rescaling was minimal and the trial averaging relevant.

4.3. Overall analysis

The results demonstrated a reasonably high overall agreement between the hip joint moments and power during the gait cycle. The magnitude of the differences and RMSEs were similar for each component of the moment along the three axes. The RMSEs of the moments appeared slightly smaller during the swing than the support phases. Furthermore, the maximum RMSEs between both inputs for moments were reasonably small and comparable to those presented in Dumas et al (2009) for the knee joint moments [22]. This might be due to the facts that the soft tissue of the residuum was well contained within the socket, and the deformation of the prosthetic foot and the shoe was limited.

4.4. Phases analysis

The overall small RMSEs in moments were translated into differences in values of peak power and total work that were more or less noticeable for each phasis.

The magnitudes of peak power were the largest during the phases H1 of generation and H2 of absorption. All combined, the duration of these two phases represented approximately 54% of the gait cycle coinciding mainly with the support phase.

However, the difference between both inputs for the phase H1 was inconsistently positive or negative. The differences for the phase H2 was only consistent for all the trials for the total work. This inconstancy might illustrate the sensitivity of the inverse dynamics to the computation of ground reactions including determination of the location of centre of pressure as well as knee and hip joint centres.

The magnitudes of peak power were smaller during the phases H3 of generation and H4 of absorption. The peak values H3 and H4 occurred during the first part of the swing. However, the duration of these two phases represented approximately 39% of the gait cycle taking place mainly during the swing phase. Interestingly, both the value of the peaks and the total work was consistently under-estimated by the inverse dynamic using ground reaction forces.

On the one hand, peak comparison must be interpreted with caution as instantaneous power is particularly sensitive to small variations in inputs. On the other hand, noticeable differences between relatively small values of peak power and total work during small percentage of one gait cycle can become potentially significant over a number of cycles, particularly in terms of energy expenditure[28-29]. Furthermore, the hip contribution around the toe-off and during first part of the swing is particularly relevant from a clinical point of view. Indeed, the hip will be partially responsible for generating sufficient knee flexion to ensure foot clearance. This is an

essential aspect in safe walking and fall prevention [30-32].
Incidentally, it should be noticed that individual curves of moments computed by inverse dynamics using knee reactions presented some “spikes” at the end of the swing phase. Those spikes were flatten by the averaging of all the trials although they appear through a slight increase in standard deviation in Figure 2. Similar singularities that occurred in previous direct measurements were attributed to the locking and terminal impact of the knee mechanism [17, 22]. Interestingly, the effect of knee mechanism on the power during H4 was minimal.
Finally, Seroussi et al (1996) relied on ground reactions to demonstrate that hip joint power on prosthetic limb presented increased of positive works during H3 compared to able-bodied [27]. The present results suggested that this works may be even larger when using knee reactions.

4.5. Contributions

This study partially validates inverse dynamics method based on ground reactions obtained with force-plates. The effects of this method’s limitations (i.e., rigid segments linked by ideal joints, sensibility to inertial parameters, time derivatives, location of joint centres and centre of pressure, and error propagation) [12-15] appeared to be more noticeable when considering peaks of power during phases of generation H3 and absorption H4 in the first part of the swing phase.
The results also demonstrate the capacity of the inverse dynamics method based on knee reactions obtained with transducer to assess correctly hip joint moments, power and work. The true accuracy of this method compared to the one based on ground reactions is difficult to determine. Nonetheless, this study confirms that it is possible to assess hip joint dynamics while depending only on experimental constraints associated with a fixed or wearable 3D motion capture systems (e.g., number of cameras, field of view)[16], as a transducer can provide dynamic data for an unlimited number of steps and activities[17, 20-21]. Therefore, the proposed method has the potential to provide more realistic hip dynamics in terms of variability over gait cycles and a wide range of activities of daily living [20-21].

4.6. Future studies

One way to strengthen the comparison of both methods would be to look at the velocity and acceleration of the centre of mass of the whole body obtained with kinematic data, ground reactions and knee reactions [33-34]. However, previous preliminary studies demonstrated that such comparison might be difficult because the velocity of the centre of mass could only be calculated during a short period of time when all the external forces were only applied to the prosthetic leg (i.e., during single support) [35].
The possibilities for longitudinal studies using inverse dynamics based on direct measurements of knee reactions are endless, particularly for the ones focusing on hip during rehabilitation [18, 23], multiple steps of walking [21] and activities of daily living [19] for a larger cohort of transfemoral amputees [20-21, 36-39] using fixed or wearable motion capture systems [16]. These studies will provide insight into ways lower limb amputees use hip joint either to stabilize or to propel during locomotion [40-41]. Studying a group of transfemoral amputees fitted with an osseointegrated fixation will also be particularly relevant since several studies demonstrated that one of the main prosthetic benefits of this type of fixation is to improve hip range of movement in absence of socket [42-44]. This will enable to determine to which extend the increase of range of movement in translated into hip joint kinetics. Further cross-sectional studies could compare prostheses constructions (e.g., residuum length, socket design, components characteristics, alignment) and record complementary clinical information such as lateral trunk bending, sound limb dynamics [1], EMG of the hip and residuum muscles [45-47] or metabolic energy consumption [6, 29, 48].

5. CONCLUSIONS

This study highlighted the difficulty of assessing dynamics at the hip joint of transfemoral amputees using conventional experimental setups relying on fixed motion capture system and force-plates. The results demonstrated that inverse dynamic calculation using ground reactions obtained with floor-mounted forces-plates and knee reactions obtained with a transducer produced comparable results. Therefore, this study confirmed that a transducer mounted within the prosthesis has the capacity to provide more realistic kinetics of the prosthetic limb for multiple steps and a wide range of activities without issues of foot placement on force-plates. It is anticipated that this proposed method is a stepping stone into kinetics of prosthetic limbs obtained with data from both wearable motion sensors and force transducers.

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REFERENCES

1. Nolan L, Lees A. The functional demands on the intact limb during walking for active trans-femoral and trans-tibial amputees. Prosthetics and Orthotics International. 2000;24(2):117-25.
2. Sjodahl C, Jarnlo G, Soderberg B, Persson B. Kinematic and kinetic gait analysis in the sagittal plane of trans-femoral amputees before and after special gait re-education. Prosthetics and Orthotics International. 2002;26(2):101 - 12.
3. Sjodahl C, Jarnlo G-B, Soderberg B, Persson BM. Pelvic motion in trans-femoral amputees in the frontal and transverse plane before and after special gait re-education. Prosthetics and Orthotics International. 2003;27(3):227 - 37.
4. Yang L, Solomonidis SE, Spence WD, Paul JP. The influence of limb alignment on the gait of above-knee amputees. Journal of Biomechanics. 1991;24(11):981-97.
5. Pitkin M. Effects of design variants in lower-limb prostheses on gait synergy. Journal of Prosthetics and Orthotics 1997;9(3):113-22.
6. Schmalz T, Blumentritt S, Jarasch R. Energy expenditure and biomechanical characteristics of lower limb amputee gait: The influence of prosthetic alignment and different prosthetic components. Gait and Posture. 2002;16(3):255-63.
7. Wühr J, Veltmann U, Linkemeyer L, Drerup B, Wetz H. Influence of modern above-knee prostheses on the biomechanics of gait. Advances in Medical Engineering Springer Berlin Heidelberg; 2007. p. 267-72.
8. Blumentritt S, Scherer H, Michael J, Schmalz T. Transfemoral amputees walking on a rotary hydraulic prosthetic knee mechanism: A preliminary Report. Journal of Prosthetics and Orthotics. 1998;10(3):61-70.
9. Segal A, Orendurff M, Klute G, McDowell M, Pecoraro J, Shofer J, Czerniecki J. Kinematic and kinetic comparisons of transfemoral amputee gait using C-Leg® and Mauch SNS® prosthetic knees. Journal of Rehabilitation Research & Development. 2006;43(7):857-70.
10. McNealy LL, Gard SA. Effect of prosthetic ankle units on the gait of persons with bilateral trans-femoral amputations. Prosthetics and Orthotics International. 2008;32(1):111 - 26.
11. van der Linden ML, Solomonidis SE, Spence WD, Li N, Paul JP. A methodology for studying the effects of various types of prosthetic feet on the biomechanics of trans-femoral amputee gait. Journal of Biomechanics. 1999;32(9):877-89.
12. Holden P, Stanhope S. The effect of variation in knee center location estimates on net knee joint moments. Gait and Posture. 1998;7:1-6.
13. McCaw S, DeVita P. Errors in alignment of the center of pressure and foot coordinates affect predicted lower extermity torques. Journal of Biomechanics. 1995;25(8):985-8.
14. Rao G, Amarantini D, Berton E, Favier D. Influence of body segments' parameters estimation models on inverse dynamics solutions during gait. Journal of Biomechanics. 2006;2006(39):1531-6.

15. Stagni R, Leardini A, Cappozzo A, Benedetti M, Cappello A. Effects of hip joint centre mislocation on gait analysis results. Journal of Biomechanics. 2000;33:1479-87.
16. Favre J, Crevoisier X, Jolles BM, Aminian K. Evaluation of a mixed approach combining stationary and wearable systems to monitor gait over long distance. Journal of Biomechanics. 2010;43(11):2196-202.
17. Frossard L, Beck J, Dillon M, Chappell M, Evans JH. Development and preliminary testing of a device for the direct measurement of forces and moments in the prosthetic limb of transfemoral amputees during activities of daily living. Journal of Prosthetics and Orthotics. 2003;15(4):135-42.
18. Frossard L, Lee Gow D, Hagberg K, Cairns N, Contoyannis B, Gray S, Brånemark R, Pearcy M. Apparatus for monitoring load bearing rehabilitation exercises of a transfemoral amputee fitted with an osseointegrated fixation: A proof-of-concept study. Gait and Posture. 2010;31(2):223-8.
19. Frossard L, Stevenson N, Sullivan J, Uden M, Pearcy M. Categorization of activities of daily living of lower limb amputees during short-term use of a portable kinetic recording system: A preliminary study. Journal of Prosthetics and Orthotics. 2011;23(1):2-11.
20. Lee W, Frossard L, Hagberg K, Haggstrom E, Brånemark R. Kinetics Analysis of Transfemoral Amputees fitted with Osseointegrated Fixation Performing Common Activities of Daily Living. Clinical Biomechanics. 2007;22(6):665-73.
21. Lee WC, Frossard LA, Hagberg K, Haggstrom E, Gow DL, Gray S, Branemark R. Magnitude and variability of loading on the osseointegrated implant of transfemoral amputees during walking. Medical Engineering & Physics. 2008;30:825-33.
22. Dumas R, Cheze L, Frossard L. Loading applied on prosthetic knee of transfemoral amputee: Comparison of inverse dynamics and direct measurements. Gait and Posture. 2009;30(4):560-2.
23. Frossard L, Hagberg K, Haggstrom E, Branemark R. Load-relief of walking aids on osseointegrated fixation: Instrument for evidence-based practice. Neural Systems and Rehabilitation Engineering, IEEE Transactions on. 2009;17(1):9-14.
24. Ehrig RM, Taylor WR, Duda GN, Heller MO. A survey of formal methods for determining functional joint axes. Journal of Biomechanics. 2007;40(10):2150-7.
25. Dumas R, Aissaoui R, de Guise J. A 3D generic inverse dynamic method using wrench notation and quaternion algebra. Computer Methods in Biomechanics and Biomedical Engineering. 2004;7:159-66.
26. Winter DA, Sienko SE. Biomechanics of below-knee amputee gait. Journal of Biomechanics. 1988;21(5):361-7.
27. Seroussi RE, Gitter A, Czerniecki JM, Weaver K. Mechanical work adaptations of above knee amputee ambulation. Archives of Physical Medicine and Rehabilitation. 1996;77:1209-14.
28. Foerster S, Bagley A, Dan Mote C, Harry B, Skinner B. The prediction of metabolic energy expenditure during gait from mechanical energy of the limb-A preliminary study. Journal of Rehabilitation Research & Development. 1995;32(2):128-34.
29. Hagberg K, Häggström E, Brånemark R. Physiological cost index (PCI) and walking performance in individuals with transfemoral prostheses compared to healthy controls. Disability & Rehabilitation. 2007;29(8):643-9.
30. Frossard L, Tranberg R, Haggstrom E, Pearcy M, Branemark R. Fall of a transfemoral amputee fitted with osseointegrated fixation: loading impact on residuum. Gait and Posture. 2009;30(Supplement 2):S151-S2.
31. Frossard LA. Load on osseointegrated fixation of a transfemoral amputee during a fall: Determination of the time and duration of descent. Prosthetics and Orthotics International. 2010;34(4):472-87.
32. Frossard LA, Tranberg R, Haggstrom E, Pearcy M, Brånemark R. Load on osseointegrated fixation of a transfemoral amputee during a fall: Loading, descent, impact and recovery analysis. Prosthetics

and Orthotics International. 2010;34(1):85-97.
33. Gordon D, Robertson E, Winter DA. Mechanical energy generation, absorption and transfer amongst segments during walking. Journal of Biomechanics. 1980;13(10):845-54.
34. Bobbert MF, Schamhardt HC, Nigg BM. Calculation of vertical ground reaction force estimates during running from positional data. Journal of Biomechanics. 1991;24(12):1095-105.
35. Stevenson N, Smeathers J, Evans J, Frossard L. Direct measurement as a tool for assessing the accuracy of inverse dynamics. V Australasian Biomechanics Conference (ABC5); Sydney, Australia: UNSW; 2004. p. 11-2.
36. Pitkin M. On the way to total integration of prosthetic pylon with residuum. Journal of Rehabilitation Research & Development. 2009;46(3):3456-360.
37. Aschoff H, Clausen A, Hoffmeister T. [The Endo-Exo Femur Prosthesis - A new concept of bone-guided prosthetic rehabilitation following above-knee amputation]. Zeitschrift der Unfallchirurg. 2009;147:610-5.
38. Pendegrass CJ, Goodship AE, Blunn GW. Development of a soft tissue seal around bone-anchored transcutaneous amputation prostheses. Biomaterials. 2006;27(23):4183-91.
39. Brånemark R, Brånemark P-I, Rydevik B, Myers R. Osseointegration in skeletal reconstruction and rehabilitation: A review. Journal of Rehabilitation Research & Development. 2001;38(2):175-81.
40. Dumas R, Cheze L. Hip and knee joints are more stabilized than driven during the stance phase of gait: An analysis of the 3D angle between joint moment and joint angular velocity. Gait & Posture. 2008;28:243-50.
41. Samson W, Desroches G, Cheze L, Dumas R. 3D joint dynamics analysis of healthy children's gait. Journal of Biomechanics. 2009;42:2447-53.
42. Tranberg R, Zügner R, Kärrholm J. Improvements in hip- and pelvic motion for patients with osseointegrated trans-femoral prostheses. Gait & Posture. 2011;33(2):165-8.
43. Hagberg K, Haggstrom E, Uden M, Brånemark R. Socket versus bone-anchored trans-femoral prostheses: hip range of motion and sitting comfort. Prosthetics and Orthotics International. 2005;29(2):153-63.
44. Sullivan J, Uden M, Robinson K, Sooriakumaran S. Rehabilitation of the trans-femoral amputee with an osseointegrated prosthesis: the United Kingdom experience. Prosthetics and Orthotics International. 2003;27:114-20.
45. Hong JH, Mun MS. Relationship between socket pressure and EMG of two muscles in trans-femoral stumps during gait. Prosthetics and Orthotics International. 2005;29(1):59 - 72.
46. Jaegers S, Arendzen J, de Jongh H. Changes in hip muscles after above-knee amputation. Clinical Orthopaedics & Related Research. 1995;319:276-84.
47. Pantall A, Durham S, Ewins D. Variability of SEMG of five stump muscles during stance phase of gait in TF amputees with osseointegrated prostheses. Gait & Posture. 2006;24(Supplement 2):S153-S4.
48. Perry J, Burnfield JM, Newsam CJ, Conley P. Energy expenditure and gait characteristics of a bilateral amputee walking with C-leg prostheses compared with stubby and conventional articulating prostheses. Archives of Physical Medicine and Rehabilitation. 2004;85(10):1711-7.

LIST OF TABLES AND FIGURES

Table 1: Mean time and value of peak power as well as mechanical work at the hip as determined by inverse dynamics using ground reactions obtained with force-plates or knee reactions obtained with transducer during typical phases of generation (i.e., H1, H3) and absorption (i.e., H2, H4) during gait cycle (GC). Positive and negative values corresponded to generation and absorption of energy, and positive and negative mechanical work, respectively.

		Phasis of power			
		H1	**H2**	**H3**	**H4**
		Mean±SD	**Mean±SD**	**Mean±SD**	**Mean±SD**
General descriptors					
Beginning	(%GC)	1.38±0.74	36.88±1.46	57.25±0.46	73.88±1.73
End	(%GC)	35.88±1.46	56.25±0.46	72.88±1.73	98±0
Duration	(%GC)	34.5±1.41	19.38±1.19	15.63±1.51	24.13±1.73
Ground reactions (Force-plates)					
Power					
Time of peak value	(%GC)	12.38±3.66	50.88±0.99	63.63±3.85	77.75±2.76
Peak value	(W)	59.17±7.5	-56.97±10.45	18.82±4.17	-5.41±0.83
Work	(J)	1270.68±231.7	-638.84±177.45	167.05±45.12	-47.95±11.39
Knee reactions (Transducer)					
Power					
Time of peak value	(%GC)	12.13±3.14	51.13±0.64	63.38±0.92	75.5±1.77
Peak value	(W)	57.47±9.89	-56.34±5.26	26.08±4.97	-11.78±1.79
Work	(J)	1286.04±263.14	-511.91±142.24	250.86±68.35	-115.54±17.9

Figure 1. Schematic representation of the components of the prosthetic leg (A: socket, B: spherical plate, C: transducer, D: pyramidal connector, E: knee, F: ankle), the marker set and alignment, the position and orientation of the socket (S[ML_S , AP_S, VT_S]) and the transducer (T[X_T, Y_T, Z_T]) coordinate systems, and angle correction to re-align both coordinates systems (θ and ω) on the sagittal and frontal planes.

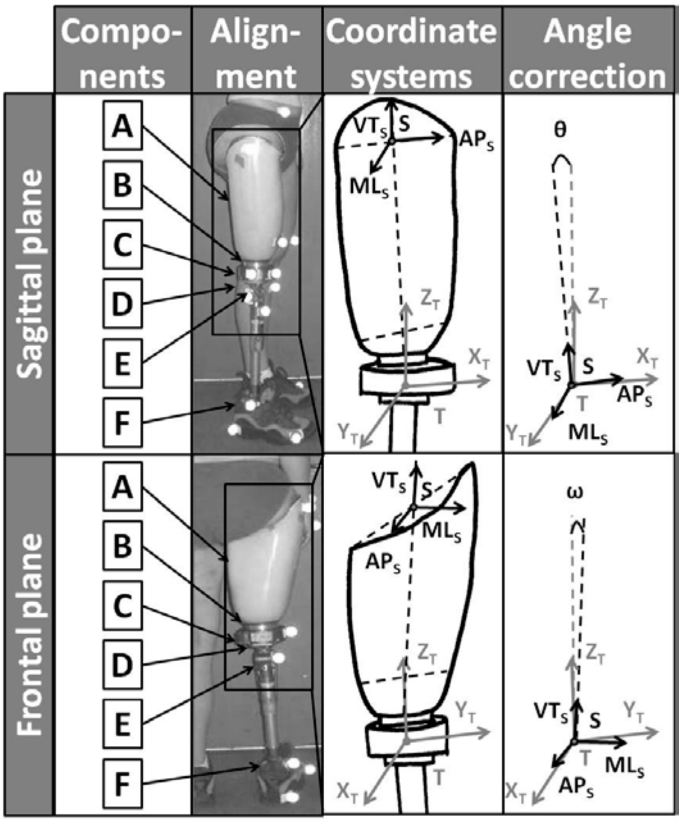


Figure 2. Mean hip joint moments in frontal, axial and sagittal planes of the pelvis coordinate system as determined by 3D inverse dynamics using ground reactions at the centre of pressure level or knee reactions at joint centre level during a gait cycle (HC: heel contact, TO: mean toe-off).

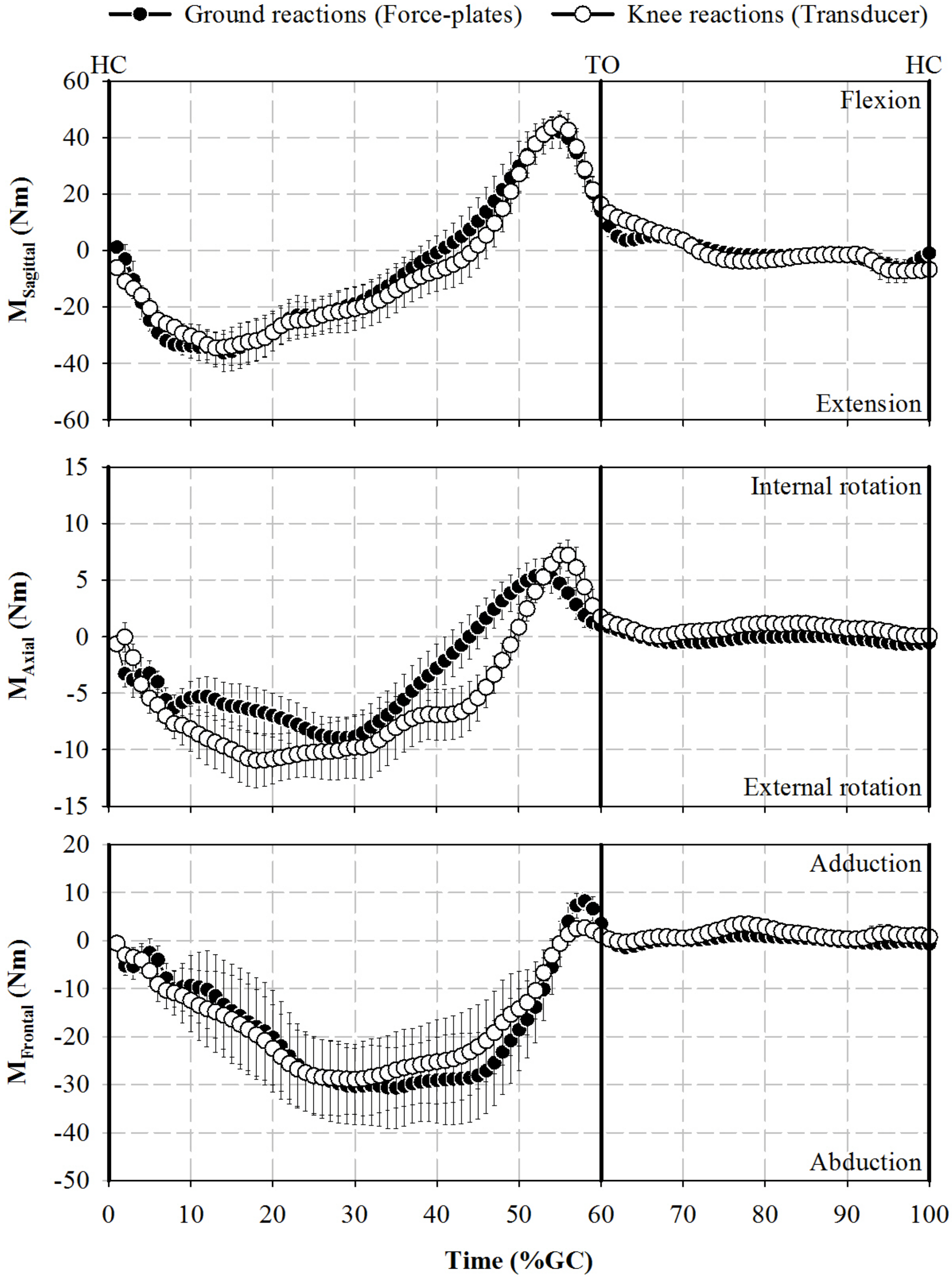
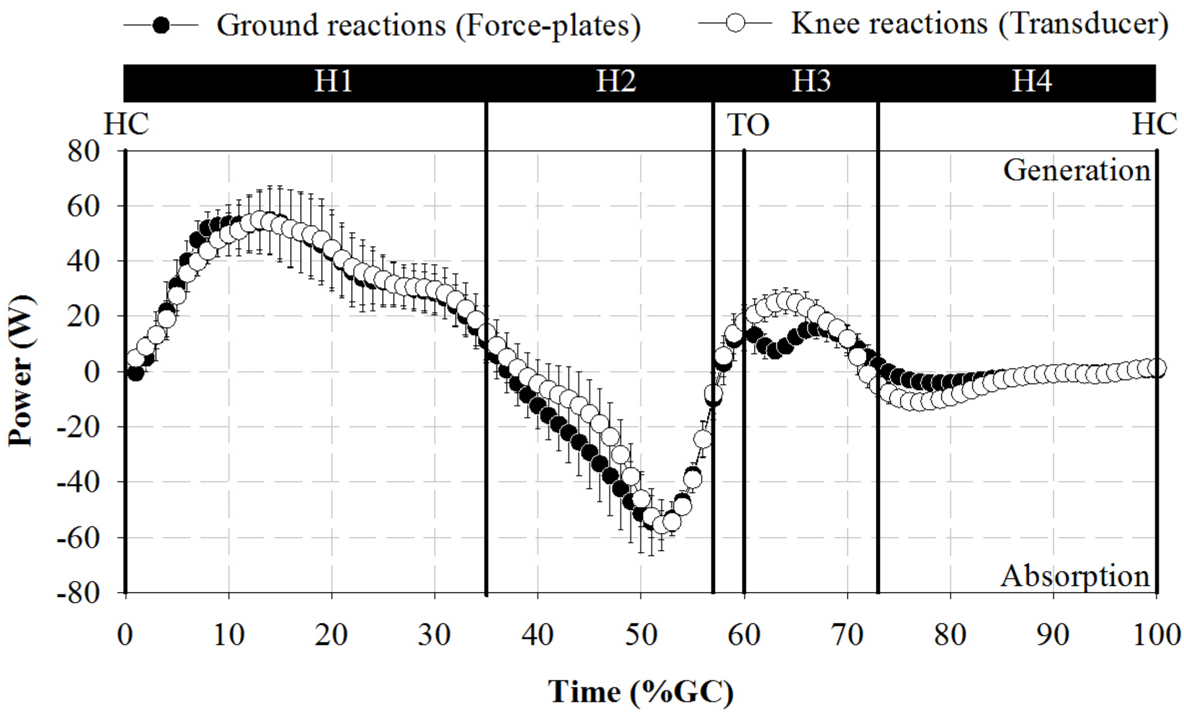


Figure 3. Mean curves and typical phases of hip power H1, H2, H3 and H4 as determined by inverse dynamics using ground reactions at the centre of pressure level or knee reactions at joint centre level during a gait cycle (HC: heel contact, TO: mean toe-off). Positive and negative values corresponded to generation and absorption of energy, respectively.



Ground reactions (Force-plates)
Knee reactions (Transducer)
H1
H2
H3
H4
HC
TO
HC
Generation
Absorption
Power (W)
80
60
40
20
0
-20
-40
-60
-80
0
10
20
30
40
50
60
70
80
90
100
Time (%GC)